A NEW MODEL-BASED ECTOPIC BEAT CORRECTION ALGORITHM FOR HEART RATE VARIABILITY

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Abstract- **We propose a new ectopic beat correction algorithm for Heart Rate Variability (HRV) that is based on the Integral Pulse Frequency Modulation (IPFM) model. The model is used to infer the modulation signal that has generated the beats surrounding the ectopic beat. The ectopic beat is then removed and replaced with a normal beat at the time which least disturbs the rhythm identified with the IPFM model. We test the performance of the technique by removing simulated ectopy from real heart rate data from human subjects using a number of previously published correction techniques as well as our own. We show that our new technique performs significantly better than existing techniques for low and high levels of ectopy.**

Keywords: **HRV, ectopic beat removal, IPFM model.**

I. INTRODUCTION

Heart rate variability (HRV) centers on the study of the autonomic nervous system via the information contained in the intervals between heartbeats. Ectopic focii are regions in the heart that produce spontaneous heartbeats, known as ectopic beats, which serve to contaminate the information contained in the heartbeats. A number of studies have shown that analyzing data containing even single ectopic beats can lead to significant over-estimation of time- and frequency-domain HRV indices [1-4]. Unfortunately, elevated levels of ectopic activity often accompany many of the conditions for which HRV analysis is potentially most useful [3, 5, 6]. Researchers often deal with ectopic beats by avoiding them. However, this is not always possible due to data length considerations and the experimental nature of many studies [2]. Also, the occurrence of ectopic beats might well be related to the nervous system [7] and therefore the exclusion of ectopic segments may bias measurements [3].

Alternative techniques attempt to mathematically remove the artifact left by the ectopic beat. Reference [3] presents a detailed comparison of the effectiveness of several popular correction algorithms. The techniques were: null (no correction), linear and cubic spline interpolation, nonlinear interpolation and simple deletion. Of these techniques, nonlinear interpolation and simple deletion had the best overall performance. Ironically, the most common correction techniques in the literature are linear and cubic splines, which reduce the distortion, but are not as effective as simply deleting the ectopic beats. None of these techniques gave reliable frequency-domain measurements during ectopic episodes.

In this paper we develop a new model based technique for the replacement of ectopic beats. It is based on the Integral Pulse Frequency Modulation (IPFM) model, which is a model of sino-atrial modulation by the nervous system [8]. We employ the IPFM model to characterize the modulation of heart rate around the time of the ectopic. The ectopic beat is then deleted and a replacement beat inserted at the time that least disturbs the rhythm identified by the IPFM model. Our technique employs a cost function minimization approach. We compare the performance of our technique to other correction techniques on real HRV data with simulated ectopy.

II. THE NEW METHOD

Our correction method is based on the IPFM model, which we describe before continuing.

A. The IPFM model

The Integral Pulse Frequency Modulation (IPFM) model was first suggested as a model for pacemaker activity by Hyndman and Mohn [8]. The IPFM model generates heartbeats by integrating an input signal until it reaches a preset threshold. At this point a pulse is produced and the integrator is reset to zero. The time of occurrence of the heartbeats are denoted by the sequence $\{t_k\}$ and for IPFM generated data satisfy the relationship

$$\int_{t_k}^{t_{k+1}} \left(1+m(t)\right)dt = \overline{I} \qquad (1)$$

where $m(t)$ is the zero-mean modulation signal and $\overline{I}$ is the mean interval between heartbeats and also the threshold for the integrate-to-threshold process. The time course of the events is represented by a series of delta functions erected at the times of the heartbeats and is denoted by $x(t)$:

$$x(t) = \sum_{k=-\infty}^{\infty} \delta\left(t - t_k\right). \qquad (2)$$

The spectrum of this function has been analytically determined by [9]. Their results show that under limited conditions the modulation signal can be recovered by low-pass filtering $x(t)$. The conditions are:

C1: $|m(t)| << 1$, i.e., small amplitude modulation.

C2: the highest frequency in $m(t)$ is less than $1/2\overline{I}$ Hz.

If an ideal low-pass filter is employed with cut-off frequency of f_c Hz, then the recovered modulation signal is the low-pass filtered event series [10], denoted by $lpfes(t)$:

$$lpfes(t) = x(t) * \mathrm{sinc}(2f_c t) = \sum_{k=1}^{N} \mathrm{sinc}(2f_c(t-t_k)). \qquad (3)$$

We take the value $f_c = 1/2\overline{I}$. If conditions C1 and C2 are satisfied, $lpfes(t) \simeq 1 + m(t)$. The low-pass filtered event

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Abstract

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series given by (3) is used to characterize the modulation of the heartbeats around the ectopic.

B. Ectopic beat replacement

Let the beat times be the sequence $\tau = \{t_k\}$ and beat number e is due to an ectopic contraction. Therefore, t_e represents the time of the ectopic contraction and is not the time that the e'th beat would have occurred had normal sinus impulse conduction occurred. We denote the perfectly corrected beat times by an asterisk superscript. Accordingly, t_e^* denotes the time of the e'th beat had the ectopic not occurred, and the sequence $\tau^* = \{t_k^*\}$ represents the sequence of beat times without ectopy.

The beat times before the ectopic contraction are identical for τ and τ^*. If the ectopic beat is assumed to be *compensatory* then the beat times after the ectopic are identical to those that would have occured had the ectopic beat not occurred. Then, τ is equivalent to τ^* except for their e'th elements. Fig. 1 depicts this scenario. Thus, the problem at hand is given τ find τ^* by replacing t_e with t_e^*. We now describe our method of accomplishing this. Given a sequence τ, we characterize the modulation with the low-pass filtered event series.

$$lpfes_\tau(t) = \sum_{t_k \in \tau} \mathrm{sinc}(2 f_c (t - t_k)) \tag{4}$$

We then define the sequence $\Psi = \{S_k\}$ where

$$S_k = \int_{t_k}^{t_{k+1}} lpfes_\tau(t)\,dt\,. \tag{5}$$

This process is depicted in Fig. 2. In addition to Ψ, the sequence $\Psi^* = \{S_k^*\}$ is defined similarly except the sequence τ^* is used instead of τ. The next result states a special property of Ψ^*:

Lemma 1: If τ^* is the output of an IPFM process with zero-mean modulation signal $m(t)$ and threshold $\overline{I}$ that satisfy C1 and C2, then Ψ^* is a constant with value $\overline{I}$.

Proof. If C1 and C2 are satisfied then $lpfes_{\tau^*}(t) = 1 + m(t)$ and by (1) the elements of Ψ^* are equal to the constant $\overline{I}$.

This result provides the means of identifying the corrected value of t_e as Ψ is constant only when $t_e = t_e^*$. As t_e must lie between t_{e-1} and t_{e+1}, we can find t_e^* by varying t_e across this range until Ψ is constant as shown in Fig. 3.

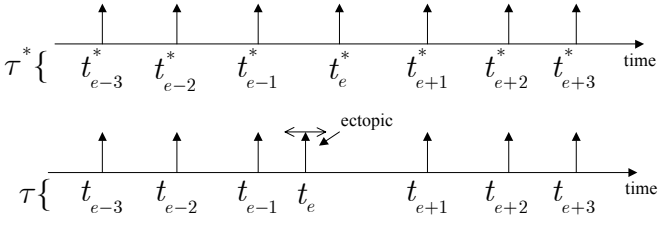

Fig. 1: The ectopic beat replacement scenario.

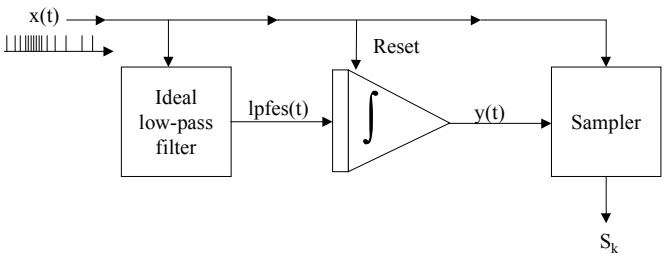

Fig. 2: The process that generates $\Psi = \{S_k\}$ from $x(t)$.

However, real heartbeats are not the result of a perfect IPFM process and Ψ may never become constant. We propose that the best estimate to t_e^* is when Ψ is "most constant". If the concept of "most constant" is defined by the sum of the squared deviations from the mean of Ψ the solution is a cost-function minimization problem with the following cost-function:

$$C(t_e) = \sum_{S_k \in \Psi} \left(S_k - E[\Psi]\right)^2\,. \tag{6}$$

The parameter t_e that gives the minimum of the cost-function is the best approximation to the true value of t_e^* and we denote it as $\hat{t}_e^*$:

$$\hat{t}_e^* = \min_{t_{e-1} < t_e < t_{e+1}} \arg C(t_e)\,. \tag{7}$$

Fig. 3 shows an example scenario. It is possible to prove that for an IPFM process with modulation signal satisfying C1 and C2, (7) provides perfect recovery, i.e. $\hat{t}_e^* = t_e^*$.

C. Efficient cost function evaluation.

The time taken by numeric minimization algorithms is fundamentally limited by the time taken to evaluate the cost function. The S_k values amount to integrating a number of $\mathrm{sinc}(\cdot)$ functions from t_k to t_{k+1}. Only the $\mathrm{sinc}(\cdot)$ functions with main lobe located close to the integration interval need to be accounted for, therefore, S_k can be approximated accurately by

$$\tilde{S}_k = \int_{t_k}^{t_{k+1}} \left[\sum_{j=k-M}^{k+1+M} \mathrm{sinc}(2 f_c (t - t_j)) \right] dt \tag{8}$$

The number of $\mathrm{sinc}(\cdot)$ functions either side of the interval is a variable parameter M which can be chosen by the operator. Smaller values correspond to less accuracy but faster execution speed. We use a value of $M = 6$.

The integration of the $\mathrm{sinc}(\cdot)$ functions can be avoided with the use of the special function

$$Si(x) = \int_0^x \left(\sin(t)/t\right) dt \tag{9}$$

The $Si(\cdot)$ function cannot be reduced to simpler analytic functions, but it can be computed efficiently by numerical means. Using (9), we can rewrite (8) as:

$$\tilde{S}_k = \frac{1}{2\pi f_c}\sum_{j=k-M}^{k+1+M} Si\left(2\pi f_c(t_{k+1}-t_j)\right) - Si\left(2\pi f_c(t_k-t_j)\right) \quad (10)$$

Equation (8) shows that $\tilde{S}_k$ is a function of t_e for $k \in [e-M-1, e+M]$. Accordingly, the set of $\tilde{S}_k$ needed to be evaluated can be restricted to this set. Also, $E[\Psi]$ can be approximated by:

$$E[\Psi] \simeq E[\tilde{S}_k] = \frac{1}{2M+2}\sum_{k=e-M-1}^{e+M} \tilde{S}_k \,. \quad (11)$$

Ectopic Replacement Algorithm

Given the events times $\tau = \{t_k\}$ *and an element* $t_e \in \tau$ *:*

1. *Discard the value of* t_e *.*
2. *From the subset* $\{t_k \,|\, k \in [e-2M-1, e+2M+1]\}$ *with* $M > 0$ *, determine* $\hat{t}_e^*$ *, the minimum of* (6)*, using* (10) *for* $\tilde{S}_k$ *and* (11) *for* $E[\tilde{\Psi}]$ *.*
3. *Replace* t_e *with the value* $\hat{t}_e^*$

For a sequence of heartbeats with multiple ectopics, the above algorithm can be applied iteratively if the ectopics are sufficiently separated. If adjacent ectopics need to be corrected for, a higher dimensional cost function minimization may be formulated in which the times of both ectopic beats are variables of the cost function. Our present analysis assumes that consecutive ectopic beats do not occur.

III. Performance on real HRV data

A. Data acquisition and ectopy simulation

Records from six healthy subjects aged between 20 and 40 years who were in sinus rhythm were manually checked and 6 five minute ectopy free segments were selected, one from each subject. For each of the 6 records, 10 separate test records were created at 2 different levels of ectopy in the same manner as described in [3]. The two levels of ectopy correspond to 12 and 120 ectopic beats per hour.

B. Ectopic correction algorithms

In addition to our new algorithm, we evaluate the performance of three existing correction techniques: linear spline interpolation, cublic spline interpolation and simple deletion. We implemented these algorithms as per [3].

C. HRV Indices

We assess each technique's ability to correct both time and frequency domain HRV indices. The time-domain indices investigated are the mean RR interval, the standard deviation of the RR intervals (SDRR), and the standard deviation of the successive differences between adjacent intervals (RMSSD). The frequency-domain indices are low-frequency power (P_L), high-frequency power (P_H) and total power (P_{TOT}). The interval spectrum is calculated by the spectrum-of-counts technique [11]. Power ranges were P_L: 0.04-0.15 Hz, P_H: 0.15-0.40 Hz and P_{TOT}: 0.01-0.40 Hz as per the recommendations in [12].

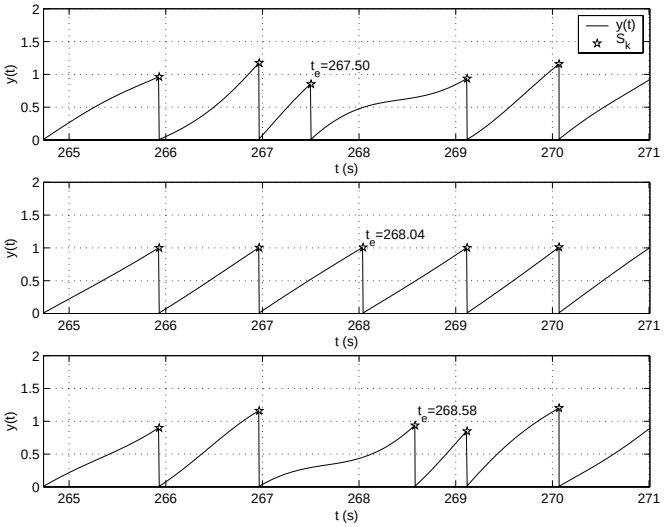

Fig. 3: The output of the process generating S_k for several values of t_e .

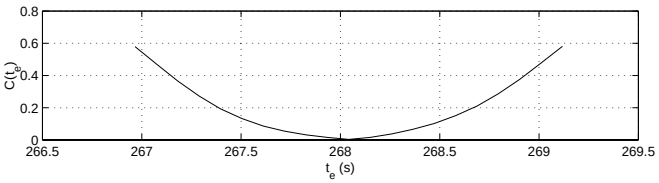

Fig. 4: The cost function $C(t_e)$. The minimum is at $\hat{t}_e^* = 268.04$ which corresponds to the most constant sequence $\Psi = \{S_k\}$ (see Fig. 3).

D. Results

In Fig. 5 and 6, the error is assessed by the root-mean-square (RMS) error, where the error is defined to be the differences between the control and the corrected values of the HRV index of interest and normalized as percentages of the mean control variable. An RMS error above 5% is deemed a clinically important departure from the control value.

The results for 12 ectopic beats per hour are summarized in Fig. 5. Our new technique is the first bar in each grouping. All techniques have an RMS error of less than 5% so there are no clinically significant errors at this level of ectopy. Our technique outperforms all the correction techniques for all HRV indices. In particular, our technique performs exceptionally well for the frequency domain measures with RMS errors of less than 0.5%. In contrast, all the other correction techniques have RMS errors of above 2% for frequency domain measures.

Fig. 6 displays the results for 120 ectopic beats per hour, which is a high level of ectopy. Even at this level, the maximum RMS error using our technique is clinically insignificant (<4% RMS error). The time domain HRV indices perform well, with clinically significant errors only occurring for the linear and cubic spline correction techniques applied to the RMSSD measure. Deletion performs almost as well as our technique. On the other hand,

for the frequency domain measures only our technique and the total power (P_{TOT}) corrected with the deletion technique have RMS errors less than the clinically significant threshold of 5%.

IV. Conclusion

In this paper, we developed and carefully analyzed the performance of a new technique for ectopy correction. The technique is based on the premise that heartbeats are generated by the IPFM model and ectopic beats are compensatory. It can be shown that if an ideal IPFM model generates heartbeats, our technique can correct isolated ectopic beats with zero error. The fact that our technique is optimal for IPFM generated intervals is reassuring, as many correction techniques are ad-hoc solutions at best, designed for simplicity rather than precision, e.g. linear and cubic splines. We have formulated the technique with robustness in mind so that even though real HRV data is not an ideal IPFM process, our technique should still perform well.

To validate the technique and obtain a comparative measure of performance we tested our technique on real data obtained from healthy subjects and compared the results with linear interpolation, cublic spline interpolation and the deletion of the ectopic intervals. The results showed that our new technique corrected the ectopic intervals with the least error for both time-domain and frequency-domain HRV indices and for low and high levels of ectopy. In particular, our technique significantly outperforms other techniques for the frequency domain measures P_H and P_L. This is a significant result, as all other techniques considered had large errors for frequency domain indices with high ectopy.

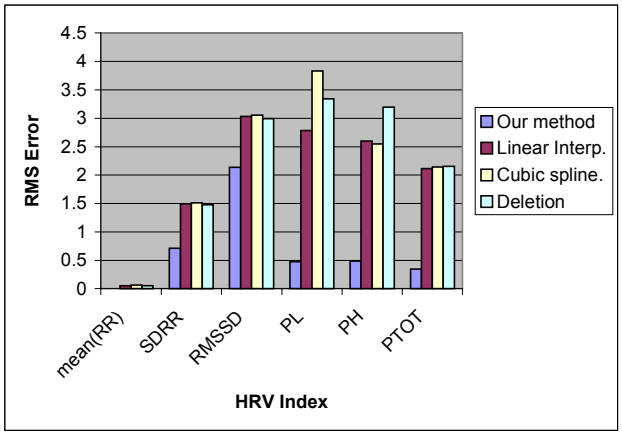

Fig. 5: 12 ectopic beats per hour.

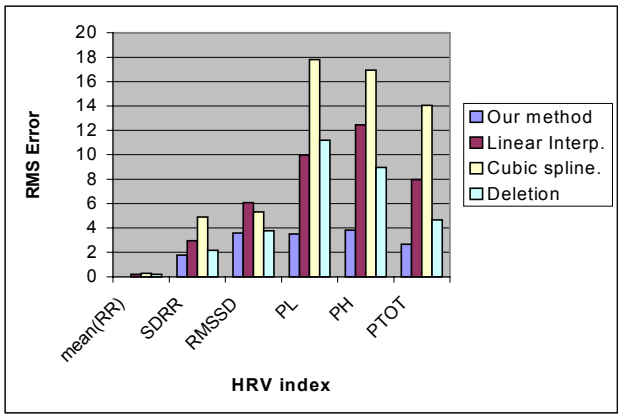

Fig. 6: 120 ectopic beats per hour.

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